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# The effects of neural synchronization and peripheral compression on the acoustic-reflex threshold<sup>a)</sup>

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This study investigates the acoustic reflex threshold (ART) dependency on stimulus phase utilizing low-level reflex audiometry [Neumann *et al.*, *Audiol. Neuro-Otol.* **1**, 359–369 (1996)]. The goal is to obtain optimal broadband stimuli for elicitation of the acoustic reflex and to obtain objective determinations of cochlear hearing loss. Three types of tone complexes with different phase characteristics were investigated: A stimulus that compensates for basilar-membrane dispersion, thus causing a large overall neural synchrony (basilar-membrane tone complex—BMTC), the temporally inversed stimulus (iBMTC), and random-phase tone complexes (rTC). The ARTs were measured in eight normal-hearing and six hearing-impaired subjects. Five different conditions of peak amplitude and stimulus repetition rate were used for each stimulus type. The results of the present study suggest that the ART is influenced by at least two different factors: (a) the degree of synchrony of neural activity across frequency, and (b) the fast-acting compression mechanism in the cochlea that is reduced in the case of a sensorineural hearing loss. The results allow a clear distinction of the two subjects groups based on the different ART for the utilized types and conditions of the stimuli. These differences might be useful for objective recruitment detection in clinical diagnostics. © 2005 Acoustical Society of America. [DOI: 10.1121/1.1867932]

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## I. INTRODUCTION

The acoustic reflex is a contraction of the middle-ear muscles induced by an intense auditory stimulus. Stimulation on either the ipsi- or the contralateral side should result in bilateral muscle contraction in a normal system. Investigations indicate that the main purpose of the reflex is to serve as an attenuator for low-frequency body noise (Simmons, 1964; Katz, 1977; Gelfand, 1998). It is believed that, of the two middle-ear muscles in humans, only the stapedius muscle contracts in response to sound as an acoustic reflex (Borg, 1973; Jerger and Northern, 1980). The reflex elicitation is normally measured acoustically by means of the middle ear's impedance change due to the middle-ear muscle contraction and hence the stiffening of the ossicular chain (Metz, 1951; Lilly, 1984). Detection of the reflex elicitation and assessment of its parameters are commonly used for clinical diagnostics of the hearing system. Deviations of the acoustic reflex threshold, for example, are used as an indicator for neural lesions affecting any portion of the reflex arc

central to the cochlea (Clemis, 1984). The pure-tone ART remains almost unaffected by sensorineural hearing loss up to 60 dB (Metz, 1951; Kawase *et al.*, 1997). Generally, the ART decreases with increasing bandwidth of the stimulus eliciting the reflex (Gorga *et al.*, 1980), similarly to the effect of loudness summation in perception. In cases of severe sensorineural or conductive hearing loss, the ART often exceeds the maximal stimulus level of 100 dB HL applied by most impedance bridges. Lower detection thresholds would be preferable, e.g., to make ART measurements usable in subjects with acute auditory damage.

The main goal of the current study was to find an optimal broadband stimulus for low-level elicitation of the ART. Therefore, we adapted a stimulus that is optimized for the measurement of auditory brainstem responses (ABR). Dau *et al.* (2000) demonstrated a significant gain of wave-V amplitude of ABR compared to click stimuli by using a phase-optimized chirp stimulus (BMchirp) that compensates for basilar-membrane travel-time differences across frequency, and thus results in a highly synchronized neural excitation. The gain of neural synchronization is reflected in higher stations of the neuronal pathway like the ventral cochlear nuclei (VCN) and the superior olivary complex (SO), where discharge timing is correlated with cochlear partition motion (Shore *et al.*, 1987; Scherg and Cramon, 1985). A stimulus

<sup>a)</sup>Parts of this study were presented at the 27th Midwinter Research Meeting of the Association for Research in Otolaryngology 2004 in Daytona Beach, Florida [Müller-Wehlau *et al.*, Abstract No. 913, p. 309].

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very similar to the chirp stimulus that was optimized for ABR measurements was tested here to reduce the ART. This seems reasonable since the afferent component of the neural pathway of the acoustic reflex can be assumed to follow almost the same path as the sources of ABR. The reflex arc comprises, among other stations, the auditory nerve (N. VIII), the VCN, and the medial nucleus of the SO, before it turns back via the facial nerve (N. VII) to the ear. Therefore, a larger excitation of certain nuclei involved in ABR measurements due to the use of phase-optimized stimulation may be accompanied by a reduction of the ART. The hypothesis tested in the present study was that the chirp stimulus suggested by Dau *et al.*, or a variant of it, may represent an ideal stimulus also for ART measurement due to the increased synchrony of the neuronal excitation. In the following, we refer to this chirp stimulus as the BMchirp. Instead of using single BMchirps, specifically designed tone complexes were used in the present study. These basilar-membrane tone complexes (BMTC) have essentially the same phase characteristics as the original BMchirps, but allow an easier analysis of the residual signal for reflex detection than the original chirps. In addition to the measurements using the BMTC stimuli, corresponding measurements were done with the temporally inverted BMTC tone complexes (iBMTC). The expectation was that the gain due to neural synchronization using the BMTC stimuli would result in a low ART, while the excitation would be highly desynchronized using the iBMTC stimuli, thus resulting in a much higher ART. As a reference, a set of noise-like stimuli was tested consisting of tone complexes with the same magnitude spectrum as the BMTC and iBMTC but with random phase components. Compared to the former stimuli these random-phase tone complexes (rTC) were expected to produce an ART that lies between those obtained with the BMTC and the iBMTC.

However, other aspects besides neural synchronization may also be important for ART determination. For example, effects of peripheral compression due to the different internal representations of the stimuli on the BM may play a role. Kubli *et al.* (2000) measured the acoustic reflex with positive and negative Schroeder-phase tone complexes (Schroeder, 1970). They explained the differences of ART for these two types of stimuli by the different internal representations at the output of cochlear filtering. The internally stronger modulated positive Schroeder-phase stimuli (S+) are supposed to be more affected by fast-acting compression on the BM—thus resulting in increased ARTs—than the negative Schroeder-phase stimuli (S−), which produce a flat internal envelope. In several psychoacoustical detection experiments (e.g., Kohlrausch and Sander, 1995; Lentz and Leek, 2001; Oxenham and Dau, 2001, 2004; Summers and Leek, 1998), the differences of internal representations produced by the Schroeder tone complexes with opposing phase have also been investigated. In these studies, modified Schroeder-phase harmonic tone complexes with different phase curvature showed a different efficiency in masking according to their different temporal modulation within the local auditory filters. These different internal representations are presumably also affected by the compressive characteristics of the BM processing and result in perceptual differences (e.g., Car-

lyon and Datta, 1997; Summers and Leek, 1998; Oxenham and Dau, 2004). A further variable affecting the ART could be the influence of temporal integration of the stimulus. Although various studies have investigated factors that act as a trigger that elicit the acoustic reflex (Kawase *et al.*, 1998; 1997; Gorga *et al.*, 1980), it is not entirely clear whether signal information is integrated within a certain time frame, or whether the peak amplitude, power, or loudness of the stimulus is appropriate to describe the internal threshold of ART elicitation. In order to test the role of temporal integration and peripheral compression in the current study, the peak-to-rms ratio was varied within a stimulus time frame of about 100 ms. In addition, experiments were carried out in normal-hearing (NH) and hearing-impaired (HI) subjects to investigate the influence of the compressive mechanisms on the BM.

## II. MEASUREMENT PARADIGMS AND DATA ANALYSIS

### A. Low-level reflex audiometry (LLRA)

To obtain improved ART measurements, i.e., low ART thresholds, we use a method suggested by Neumann *et al.* (1996), called low-level-reflex audiometry (LLRA). For tone pulses, this method is more sensitive than the conventional paradigm (Tolsdorf *et al.*, 2004). Also, the short stimulation time used in this method is more comfortable for the subjects than the stimulation used in the common method. This is especially important since the acoustical stimulation in this study was carried out with levels up to 103 dB SPL. The LLRA uses the same measurement paradigm and equipment as typically employed for the recording of otoacoustic emissions. In this method, rather than using two signals at different frequencies (the evoking stimulus and a continuous test tone—mostly at 226 Hz) as commonly used, a stimulus consisting of two identical short pulses is used to elicit and detect the reflex [see Figs. 1(a) and (d)]. The technique is based on the following principles: If the reflex is elicited during the first stimulus pulse and holds, the eardrum impedance has changed during the presentation of the second pulse. This change of impedance causes a difference between the recorded time signal of the first and second pulse within the sealed ear canal. Since the change in impedance due to the acoustic reflex has a latency of some tens of milliseconds, the second tone pulse is presented after a sufficiently long time following the first, thus leading to a maximal difference of the measured ear-canal response between these two pulses. The difference signal, or the residual of the ear-canal signal, recorded during the presentation of the two tone pulses, is analyzed to indicate the elicitation of the acoustic reflex.

Without an impedance change of the eardrum, i.e., if the first stimulus pulse elicits no reflex, the recorded ear-canal signal is almost the same during both pulses [Fig. 1(b)]. Thus, the spectrum of the difference signal mainly reflects the physiological noise and the noise of the measurement system [Fig. 1(c)]. In the case of an elicited reflex [Fig. 1(e)], the spectrum of the residual shows the frequency components of the stimulus signal [Fig. 1(f)]. The existence of these

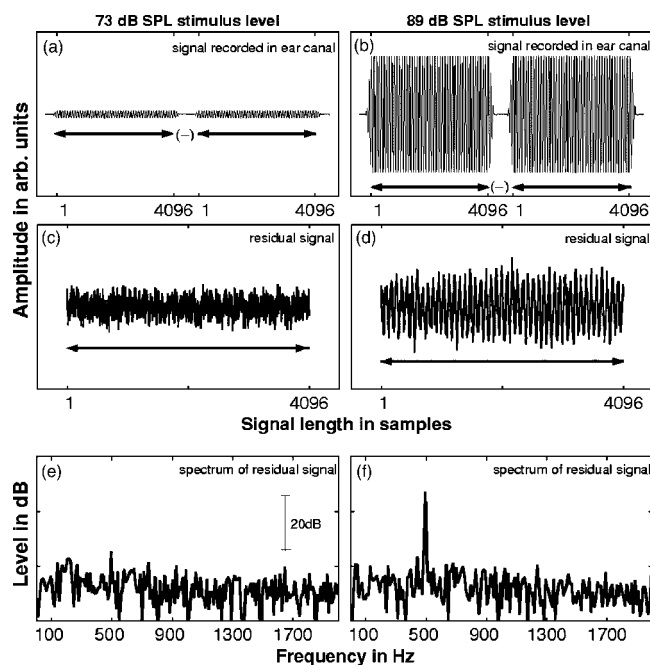


FIG. 1. Illustration of the low-level reflex audiometry (LLRA) method shown for a sinusoidal signal of 500 Hz at two different stimulus levels. The signal consists of two identical pulses of approximately 93 ms (4096 samples) duration, separated by a small gap of 10 ms, and is presented via an OAE probe. The reflex will change the middle ear's reflective properties during the presentation of the stimulus pair, thus changing the recorded signal [(a) and (d)]. By subtracting the two stimulus pulses from each other, the residual signal is obtained [panels (b) and (e)]. If the stimulus level is high enough to elicit the acoustic reflex, the residual is constituted by the stimulus signal as seen in the power spectrum (f). Otherwise the residual is dominated by noise components (g). The residual spectra are shown only for better illustration of the residual properties. The reflex elicitation is detected by means of the phase coherence of an appropriate frequency component in successive presentations.

frequency components indicates the elicited reflex.

A reliable detection of the stimulus component(s) within the residual signal is essential for the correct detection of the acoustic reflex. Further criteria are needed, especially at higher stimulation levels (close to the limit of the experimental setup), to distinguish between difference components due to the acoustic reflex and physiological or system artifacts. The low-level reflex measurement (Neumann, 1997) utilized for this study was supplemented with a different threshold criterion (see below), since the original criterion used was shown not to be sufficiently reliable (Müller-Wehlau *et al.*, 2002).

## B. Analysis methods and ART criterion

The analysis method originally suggested by Neumann *et al.* (1996) is based mainly on a signal-to-noise criterion for the frequency component(s) of the stimulus within the magnitude spectrum of the residual signal, and a further rejection criterion to account for system distortion. However, at higher stimulus levels this method sometimes indicated an ipsilateral acoustic reflex due to artifacts such as heartbeat, even in cochlear implant (CI) patients with complete hearing loss and the CI turned off (from our own unpublished data). In the current study, the threshold criterion was based on a coherence synchrony measure (CSM), which is a highly ac-

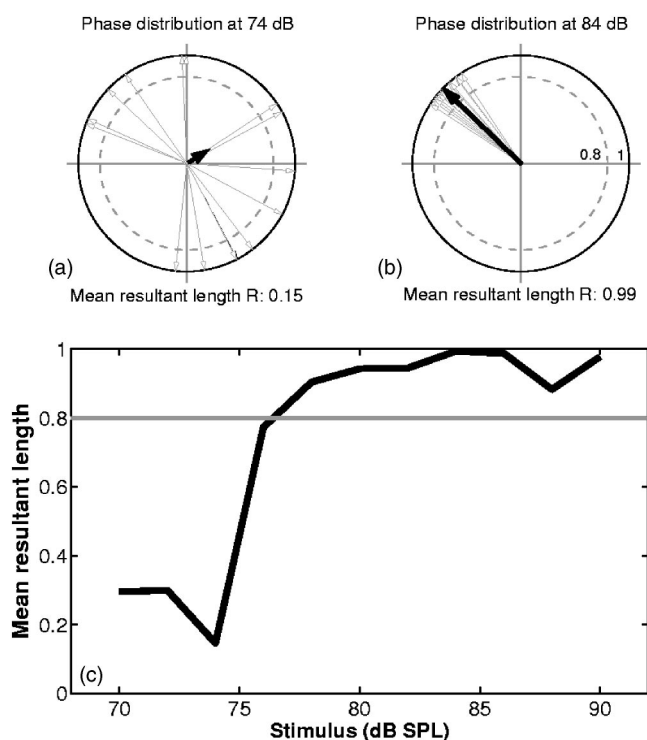


FIG. 2. Phase coherence as a measure of the reflex elicitation: For low stimulus levels the phase of the residual for a number of successive presentations (gray open arrows) is not coherent [panel (a) at 74-dB SPL stimulus level]. Thus, the mean phase vector [black arrow in the panels (a) and (b)] and hence the mean resultant length are small. If the reflex is present during the stimulation, the phase becomes more coherent [panel (b)]. The dashed circle with radius 0.8 in (a) and (b) indicates the threshold condition for the reflex detection. If the length of the mean phase vector is larger than the radius of this circle, i.e., if the mean resultant length  $R$  is larger than 0.8, the reflex is considered as elicited. The development of the mean resultant length depending on the stimulus level is shown in panel (c) with the threshold value marked by the gray line, which corresponds to the dashed circles in panels (a) and (b).

curate statistical indicator in signal detection (Valdes *et al.*, 1997). The CSM takes the reproducibility of  $n$  repetitive measurements (in this study  $n = 16$ ) as the criterion to detect the elicited reflex. The CSM is similar to the Rayleigh test of circular uniformity (Mardia, 1972) and can be considered as a measure of phase coherence calculated only from the phase values of a selected frequency component from  $n$  successive measurements without considering the amplitude of the signal spectral component. The threshold criterion is given by the mean resultant length  $R$ , i.e., the absolute value of the vectorial mean of the normalized phase vectors for a selected frequency component from  $n$  consecutive measurement intervals. This method takes into account that successive stimulations demonstrate fast stabilization of their phase values if the stimulus level is high enough to elicit the reflex, thus resulting in highly coherent phase values. This results in a small vectorial mean of the phase vectors, i.e., a small value of  $R$  [see Fig. 2(a)] if the phases from consecutive measurement intervals of the selected frequency component are randomly distributed. In contrast, similar phase values of consecutive residuals result in a value of  $R$  close to 1 [see Fig. 2(b)]. The mean resultant length  $R$  can be computed from the phase values  $\theta_i = \theta_1, \dots, \theta_n$  of the selected frequency components by



$$R = \left| \frac{1}{n} \sum_{i=1}^n r_i \right|, \text{ with } r_i = \cos \theta_i + i \sin \theta_i. \quad (1)$$

Depending on the phase coherence, the mean resultant length can take values between 0 and 1. If the resultant length is higher than (an empirically found value of)  $R \geq 0.8$ , the reflex is assumed to be elicited (see Fig. 2). This value for  $R$  is higher than those commonly used for signal detection in noise by the Rayleigh test.<sup>1</sup> This higher  $R$  value represents therefore a more conservative criterion for the reflex elicitation, and meets the fact that ambient factors give rise to small differences in the recorded microphone signal, thus resulting in the presence of spectral components even if the reflex is not present.

Additional artifact suppression was used by rejecting all single measurements whose individual residual amplitude at the selected frequency component was not within a 6-dB margin of the median of all measurements at the respective stimulus level.

The statistical evaluation in the present study was based only on the analysis at one frequency (close to 1000 Hz). A detailed examination of the evaluation frequency by using broadband stimuli showed that between 500 and 1500 Hz the reflex detection does not depend on the selected frequency component. Within this frequency band the change in middle-ear impedance is relatively large, resulting in a clear residual signal if the reflex is elicited. At lower frequencies the phase coherence is more affected by ambient low-frequency noise, while there are broad frequency bands with a strongly reduced change in impedance at higher frequencies.<sup>2</sup>

### C. Stimuli

All stimuli consisted of two identical signal frames (see Fig. 1) with frequency components between 100 and 8000 Hz and 4096 samples in length. Since the sampling frequency was 44.1 kHz, the duration,  $T$ , of a single stimulus frame was 92.88 ms. The frequency components were adjusted to the signal length, i.e., the exact frequencies were chosen to be multiples of the fast Fourier transform (FFT) base frequency,  $1/T$ . ARTs were measured for three different types of stimuli. Since our data analysis requires that an appropriate frequency component is presented during the stimulation, all signals used in the experiments were chosen as tone complexes. Twenty samples of Hanning-shaped ramps were added at the beginning and the end of each 4096-sample-long stimulus plateau. Two stimulus frames were separated by a 50-ms gap to be used as a stimulus signal by the LLRA method. Presentations of this frame pair were 1.15 s apart to allow the reflex to decay before subsequent stimulations. In optimization measurements for the LLRA, these settings were found to result in largest residual signals (from our own unpublished data).

#### 1. Tone complexes compensating for cochlear delay across frequency

These stimuli, referred to as the basilar-membrane tone complexes (BMTC) in the following, were generated by adding frequency components with phases that hypothetically

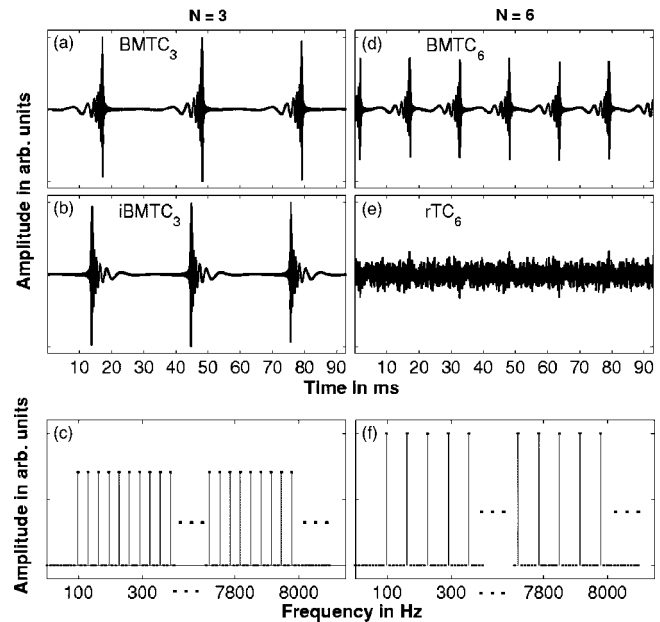


FIG. 3. Stimulus signals: BMTC<sub>3</sub> (a), iBMTC<sub>3</sub> (b), BMTC<sub>6</sub> (d), and rTC<sub>6</sub> (e). All signals are scaled to the same rms level and exhibit a flat spectral envelope with a varied number of contained frequency components. By adding frequency components that are separated by a multiple of the FFT base frequency apart, the number of chirp periods within the time frame is altered without changing the general temporal shape of the successive chirp. The amplitude spectra shown in panels (c) and (f) correspond to the stimuli shown in panels (a) and (b), and (d) and (e), respectively.

compensate for the BM travel-time differences between the different spectral components contained in the stimulus. The stimulus generation was based on the computation of the “approximate” chirp stimulus as defined in Dau *et al.* (2000) that was optimized for ABR recordings. According to Dau *et al.* (2000), the propagation time required for the calculation of the respective phase values was estimated using the cochlea model proposed by de Boer (1980) and the frequency-place transformation suggested by Greenwood (1990).

The phase of each frequency component of the tone complex was chosen as follows: The instantaneous phase,  $\varphi_{\text{inst}}$ , of the original BMchirp was calculated for the time  $t = t_{\text{fs}}$  when the instantaneous frequency of the BMchirp equals the frequency,  $f_s$ , of the selected tone complex component. The starting phase,  $\varphi_0$ , for the frequency component at time  $t = 0$  was computed such that this component has the phase  $\varphi_m$  at the time  $t = t_{\text{fs}}$ . By superimposing the components with a frequency spacing corresponding to the base frequency of the selected time frame, the respective time signal of a single chirp with flat spectral envelope is achieved [see Figs. 3(c) and (f)].

#### 2. Temporally inverted tone complexes

The second class of stimuli was generated by temporally inverting the BMTC stimuli. In the following, these stimuli are referred to as the inverted basilar-membrane tone complexes [iBMTC; see Fig. 3(b)]. The expectation was that, by

inverting the BMTC stimulus, the amount of neuronal excitation would be highly desynchronized, thus leading to an increased ART.

### 3. Random-phase tone complexes

Corresponding measurements were also obtained with a third tone complex with identical magnitude spectrum but random phases of the components, referred to as the random tone complexes [rTC; Fig. 3(e)]. The rTC stimulus for one measurement was generated with respect to one uniformly distributed random phase vector. To exclude incidental compression or synchronization effects due to this certain random phase vector, the measurements were carried out for three rTCs generated with different sets of the random phases.

### 4. Number of chirp periods per frame—frequency spacing

The BMTC and iBMTC stimuli, comprising frequency components with a spacing equal to the FFT base frequency, exhibit one chirp within the stimulating time frame. Doubling the frequency spacing gives rise to a time signal exhibiting two chirp periods within the time frame of about 100 ms. Further increase of the spacing by a factor  $N$  results in an increasing number of  $N$  “overlapping” chirps in the time domain. In the following, the number  $N$  of the chirps used in a certain stimulus is indicated by an index in the stimulus name [e.g., BMTC<sub>3</sub> for a BMTC stimulus comprising three chirps per recording frame; see Figs. 3(a), (b), and (c)]. The same notation is used for the rTC stimuli, although the recurring structure in the time domain is not as clearly seen as for the chirp stimuli.

At a fixed rms value, the number of chirp periods ( $N$ ) and hence the peak-to-rms ratio was varied [compare Figs. 3(a), (d), and (e)] in order to investigate possible summation and compression effects within one stimulus frame. The duration of the original BMchirp for the frequency range used in the current study is 10.4 ms (Dau *et al.*, 2000). We refer to this chirp length as the effective BMchirp duration. Using a maximum number of  $N=7$  successive chirps within a stimulus frame of about 100 ms avoids a significant overlap of the chirps within the effective duration. Therefore, interactions of successive chirp periods in the same BM regions within the stimulation can be mostly excluded.

### D. Detection of middle-ear muscle reflex versus detection of medial olivocochlear efferent reflex

In general, we assume that the difference of the recorded signals during the two stimulation intervals is mainly due to a contraction of the middle-ear muscle (MEM). In normal-hearing subjects it is conceivable that the residual signal is affected as well by the medial olivocochlear (MOC) efferent reflex. Thereby, the MOC reflex needs to cause a change of a stimulus frequency otoacoustic emission (SFOAE) that is elicited by the probe stimulus (Guinan *et al.*, 2003). Analogous to the difference of the two stimulation intervals due to the MEM, this would result in a residual signal. The residual signals in HI subjects should not be affected by the MOC anyway, since no or only weak SFOAE can be expected for

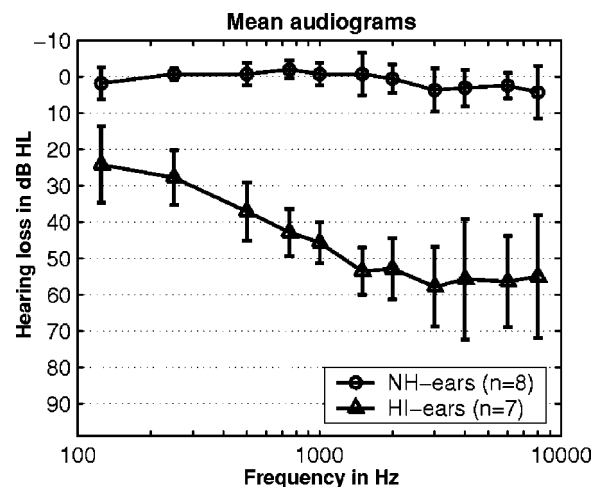


FIG. 4. Average hearing levels for NH subjects (circles) and HI subjects (triangles). The error bars represent the standard deviation of the mean thresholds across subjects.

flat hearing losses of about 50 dB. Even in the NH subjects we expect no relevant effect of the MOC reflex on the residual signal, since the stimuli used here are either noise- or chirp-like with a high sweep rate. Although these types of stimuli are appropriate to elicit the MOC reflex, both are unlikely to generate a sufficiently stable SFOAE to allow the detection of the MOC reflex in the residual signal. This holds especially for the noise-like rTC signals. Guinan *et al.* (2003) described the detection of MOC/MEM reflexes based on the change of an SFOAE evoked by a continuous sinusoid. They pointed out that, for a residual signal dominated by the MOC reflex, a rotating phase (i.e., a long group delay) is expected, as known from SFOAE, while for a MEM-dominated residual signal a short group delay can be assumed. An offline analysis of the phase characteristic of the residual signal was performed at the ART level to test for a relevant influence of the MOC on the residual signal and thus on the acoustic reflex detection. This was done for rTC and BMTC at  $N=3$  measurements in normal-hearing subjects. BMTC and rTC showed the lowest thresholds and, for  $N=3$ , the spacing of the frequency components is sufficiently close (ca. 30 Hz) to allow a reliable phase analysis across frequency.

## III. EXPERIMENTAL METHODS

### A. Subjects

Eight normal-hearing (NH) subjects (five female, three male) aged between 23 and 32 (average 28 years) with hearing thresholds better than 15 dB HL, and six hearing-impaired (HI) subjects (four female, two male) aged between 38 and 67 (average 54 years) with flat, sensorineural hearing loss participated in this study (see Fig. 4). The NH subject group had no known history of audiological diseases.

The HI subjects were chosen under the assumption that the compressive nonlinearity on the BM will be greatly reduced in these subjects. The ARTs, in response to broadband stimulation, can be expected to be elevated to some degree depending on the hearing loss. The members of the HI sub-

ject group were restricted to subjects with a flat, moderate hearing loss of approximately 50 dB. The subject LP was measured on both sides so that for this group a total of seven measurements was performed.

An audiological examination was carried out on all subjects, including reflex audiometry with a standard impedance audiometer (Grason-Stadler GSI33). The reflex threshold was ascertained by a well-established method in order to make sure that the subjects showed ARTs below 100 dB HL. Subjects showing no conventionally measured ARTs within this range were excluded from further measurements since the experimental setup was limited to stimulus levels of 103 dB SPL. The limitation in sound levels was due to both technical reasons and the goal to restrict the exposure of the subjects to high-level sound over the estimated measurement period of up to 2 h for the full range of experiments conducted. Furthermore, subjects with tympanograms showing only small changes ( $<0.3$  ml) in compliance were also excluded since the LLRA equipment provides no pressure equalization. No abnormally large changes in compliance ( $>2$  ml) have been observed within the subject groups.

## B. Setup

The whole measurement was PC based and implemented in a customized program. The level of the signal was digitally controlled and varied on the PC. A digital I/O-card (RME DIGI 96) in the PC was used for the replay of the stimulus signal, which was transmitted via an external DA/AD converter (RME ADI 8 DS) to a headphone buffer (TDT HB6) to drive the probe speaker (Otodynamics ILO BT-type OAE probe). The signal in the ear canal was recorded with an inserted probe microphone (Otodynamics ILO BT-type OAE probe) linked via a connection box that provided the required bias voltage. The microphone signal was amplified by an external low-noise amplifier (Stanford Research SR560) and then directed to the AD converter. The microphone chain was calibrated according to Siegel (2002) using a Bruel & Kjaer type 4192 microphone capsule as reference. The output path including the probe's speakers was calibrated using an artificial ear for insert earphones (Bruel & Kjaer 4157) and a broadband (150–10 000-Hz) calibration signal with flat temporal envelope. The transfer function obtained by this calibration procedure was used to calculate a phase invariant overlap-add filter to correct the stimuli for the frequency response of the output system. No individual correction or in-the-ear calibration was performed.

Before each measurement, the fit of the OAE probe in the individual ear was tested online by presenting a broadband signal and recording with the OAE probe in the sealed ear canal. The spectrum of the recorded signal was displayed in comparison to a reference spectrum obtained in the artificial ear (Bruel & Kjaer 4157) with the same procedure. The fitting of the probe in the individual ear canal was altered to obtain a sufficient correspondence between the reference and the current spectrum.

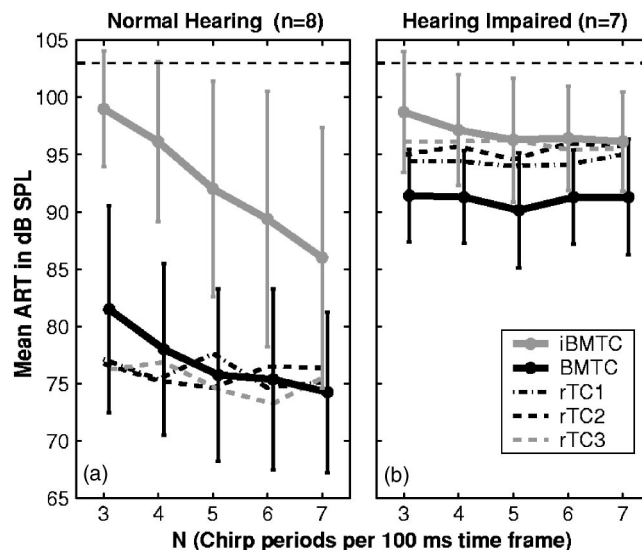


FIG. 5. Mean acoustic reflex thresholds (ART) for NH [Panel (a)] and HI listeners [Panel (b)]. The dashed horizontal line at 103 dB SPL indicates the maximal applied stimulus level. The error bars indicate the interindividual standard deviation of the respective ART measurement. For the reason of clarity, no error bars are given for the three rTC-type stimuli.

## C. Measurement

An automatic measurement mode was used to assert the reflex threshold starting at medium stimulus levels and subsequently increasing or decreasing the level depending on the reflex detection. After each reversal, the increment/decrement was reduced from 6 dB in the beginning down to 1 dB after the final reversal. Depending on the direction of the level change, either the first or the last measured point where the reflex could successfully be detected after the final reversal was taken as the acoustic reflex threshold. The automatic mode utilized a range of 50 to 103 dB SPL. No reflex threshold was recorded if the reflex could not be detected for three successive measurements at the maximal stimulus level of 103 dB SPL. Since the resulting ARTs were expected to depend on the stimulus type, the starting levels of the automatic algorithm were different for the respective stimuli. All measurements took place inside a sound-attenuating hearing booth (IAC 1203) where the subjects rested in a chair and were allowed to read. Each stimulus was presented 16 times for each of the measured presentation levels. The measurement took approximately 20 min for each of the five stimuli (each for  $N=3$  to  $N=7$ ). Therefore, all measurements in one subject were performed in a single session of about 2-h duration.

## IV. RESULTS

The results were similar within each of the two subject groups, but differed significantly between the two groups ( $p<0.005$ ).<sup>3</sup> Mean data are shown in Fig. 5.

### A. ARTs in normal-hearing subjects

The NH subjects all exhibit significantly lower ARTs for the BMTc stimuli compared to the iBMTc stimuli. The acoustic reflex thresholds of these two stimuli show a clear



dependency on the number of chirps,  $N$ , within the stimulus time frame. Paired-samples  $t$ -tests reveal the significant decrease of the ARTs with increasing  $N$  for both stimuli.<sup>4</sup> This is the case for the BMTC stimuli, where mean thresholds decrease from 81.5 to 74.3 dB SPL, and more pronounced for the iBMTC stimuli with mean thresholds dropping from 98.6 to 86 dB SPL, thus resulting in a convergence that can be generally observed in the NH group. Surprisingly, the ARTs for the rTC stimuli are equal to or even lower than those obtained for the BMTC signals. Two-sided paired-samples  $t$ -tests show significantly lower ARTs in response to the rTC stimuli for  $N=3$  chirp periods within the time frame for rTC<sub>1</sub> and rTC<sub>3</sub>, but no significant difference between any rTC stimulus and BMTC for  $N=7$ . It can also be observed that the rTC stimulus type does not show a dependency on  $N$  with the mean thresholds nearly constant around 76 dB SPL.<sup>5</sup> As expected, all three stimuli of the rTC type with different random phase vectors lead to the same ART.

The characteristics of the ARTs for the different stimulus types were similar among all NH subjects, although the absolute ART levels values for the same stimuli varied between the subjects. In some cases, differences of the ART for the respective stimuli between two NH subjects were up to 15 dB. This difference was also observed for acoustic reflex thresholds measured at 500 and 1000 Hz with a standard procedure (GSI 33 impedance audiometer). If the thresholds in response to the BMTC stimuli were elevated, it was not always possible to assert the threshold for the iBMTC signals due to the limitation of the presentation levels. This was the case for three of the eight NH subjects. In cases where the iBMTC threshold could not be determined, the ART was assumed to be 1 dB higher than the maximal tested stimulation level of 103 dB for statistical analysis. Therefore, the mean values of the iBMTC thresholds, as shown in Fig. 5, are most likely underestimated to some extent. This holds especially for stimuli comprising a low number of chirps within the stimulating time frame, where the resulting thresholds were particularly high for this stimulus type.

Due to the differences in the absolute ART levels, the interindividual standard deviation seen in Fig. 5 for the BMTC and iBMTC thresholds is quite large. Nevertheless, the key properties exhibited by this subject group, i.e. (i) the large difference between the ARTs for the BMTC and iBMTC stimuli; (ii) the dependency of these ARTs on the frequency spacing; and (iii) the low thresholds resulting from the rTC-stimuli, are the same for all subjects of the NH group.

## B. ARTs in hearing-impaired subjects

For the HI subjects (right panel of Fig. 5), the ARTs for the BMTC stimuli are also significantly lower than the ARTs obtained by stimulation with iBMTCs. However, the threshold differences between these stimuli are distinctly smaller<sup>3</sup> than for the NH subjects, and range from 8 dB for  $N=3$  to 5 dB for  $N=7$ . Second, in contrast to the NH subjects, no significant difference of the mean ARTs can be found as a function of  $N$ , either for the BMTC or the iBMTC stimulus.<sup>6</sup> The mean difference of ART between  $N=3$  and  $N=7$  for iBMTC stimuli in the HI group is only 2.5 dB (98.6 dB for

$N=3$ , and 96.1 dB for  $N=7$ ) compared to 12.6 dB in NH subjects. The mean ART in the HI subjects for the BMTC stimuli are nearly independent of  $N$  (about 91 dB SPL), while the NH subject group showed a significant<sup>3</sup> decrease of 7.2 dB with increasing  $N$ .

Even though the BMTC thresholds found in the HI subjects were elevated compared to the NH subjects, it was possible, with one exception, to assert all ARTs for the iBMTC stimuli in this subject group.

As for the NH subjects, the three rTC stimuli led to essentially the same ART, independent of the frequency spacing and the random vector used for the generation. However, the BMTC thresholds found for the HI were lower than those found for the rTC stimuli, in contrast to the results of the NH group.

For one subject (LP) of the HI group, the pure-tone hearing thresholds for the right ear were about 15 dB lower than for the left ear. A difference of the ARTs for the respective signals can be observed between the two sides, with slightly elevated thresholds for all stimuli on the worse side compared to the thresholds measured in the better ear [compare Figs. 7(a) and (b)]. It can also be observed that the threshold difference between the BMTC and the iBMTC becomes smaller and the dependency on  $N$  less pronounced, especially of the iBMTC on the worse ear.

Basic ART characteristics for all subjects are summarized in Table I as the ART  $T$  of rTC stimuli (for  $N=3$ ), the difference  $D$  between the ART from rTC and iBMTC stimuli (for  $N=3$ ), and the decrease  $G$  of ARTs for iBMTC from  $N=3$  to  $N=7$  (for illustration, see also Fig. 8).

## C. Detection of middle-ear muscle reflex versus detection of medial olivocochlear efferent reflex

In order to exclude possible effects of the MOC on the acoustic reflex detection, the phase characteristics of the residual signal across frequency were investigated. All normal-hearing subjects exhibited a constant phase across frequency at threshold levels, indicating that the residual signals are clearly dominated by the MEM contraction (Guinan *et al.*, 2003). This corresponds to the findings of Guinan *et al.* (2003), who found that for elicitor levels of 65 dB SPL or higher, the residual signal is either MEM dominated or a mixture of MEM and MOC.

Furthermore, to exclude the possible influence of spontaneous otoacoustic emission (SOAE) that might be triggered by the stimulus and thereby obscure the ART, we conducted an offline examination of the residual signal at several frequencies between 500 and 1500 Hz using the analysis method mentioned above. This examination did not show the frequency specificity that could be expected if the residual signal was caused by SOAE. All frequencies within certain bands were equally appropriate to detect the reflex, indicating that the residual signal was caused by the impedance change resulting from the MEM contraction.



TABLE I. Comparison of the individual difference  $D$  between ARTs for rTC and iBMTC stimuli and the difference  $G$  of the ARTs for iBMTC stimuli between iBMTC<sub>3</sub> and iBMTC<sub>7</sub> (see Fig. 8). The ART  $T$  of the rTC stimuli for each subject were calculated with respect to the mean ART of the three rTC type stimuli. No ART for the iBMTC at  $N=3$  could be obtained for the subjects indicated by the asterisk (\*). In these cases the ART for the iBMTC<sub>3</sub> stimulus are approximated from the slope of the remaining iBMTC thresholds. This was not possible for subject JJ, where only the iBMTC<sub>7</sub> threshold could be measured. The  $D$  value for this subject is estimated from the difference between the mean rTC thresholds and an assumed iBMTC threshold of 104 dB SPL.

NH subjects					HI subjects				
Subject	$D1$ (dB)	$D2$ (dB)	$G$ (dB)	$T$ (dB)	Subject	$D1$ (dB)	$D2$ (dB)	$G$ (dB)	$T$ (dB)
NJ	23	-4	16	68	FR	0	4	2	91
MM	34	-7	32	67	BU	2	13	-1	100
OM*	26	-4	12	78	FD	1	5	-3	92
SA	21	1	14	73	FW*	10	-2	7	94
JJ*	18	-4	...	86	WW	-1	8	2	97
KA*	18	-11	16	86	LP (better ear)	8	2	7	94
BS	16	-1	14	82	LP (worse ear)	4	2	4	99
LA	22	-9	10	74					

## V. DISCUSSION

### A. Mechanisms affecting the acoustic reflex thresholds

The idea behind the generation of the stimuli used in the present study was based on the hypothesis that the reflex threshold is related to the amount of synchronized neural excitation produced by the respective activating stimulus. The experimental results found in the current study partially support this hypothesis. The data also suggest that peripheral compression strongly influences the results for the different stimuli. In all HI subjects, the “optimized” BMTC produced the lowest ARTs [see Figs. 6(c) and (d)]. The ARTs obtained with the noise-like rTC stimuli decreased with decreasing hearing loss [Figs. 7(b) and (a)] and obtained values slightly below those for the BMTC stimuli [Figs. 6(a) and (b)] in the NH subjects. This effect and several of the other key observations in the data are discussed in the following. A detailed modeling of the effects was beyond the scope of the study. However, we have attempted to at least qualitatively explain the results based on the different aspects associated with cochlear processing.

#### 1. Excitation characteristics of the different stimuli

The BMTC were designed to compensate for BM dispersion. Ideally, these stimuli produce a maximum amount of excitation across frequency at a particular point in time. In a nonideal case, e.g., if the sweep rate of the chirp does not exactly compensate the delay line characteristic of the cochlea, a relatively broadband synchronized excitation still can be expected that moves in the apical or basal direction. The summation of excitation for the BMTC across all frequency bands as a function of time results in a peaky, i.e., temporally highly modulated “spectral summation response,” with the maximum at the time when each auditory filter reaches its maximal excitation. The BMTC are trains of up-chirps, with the instantaneous frequency of each single chirp moving from low to high frequencies. A relatively flat temporal response (slowly increasing and decreasing in time) in each *single* (local) auditory filter can be expected for up-chirps since the stimulus phase curvature has the same sign as the curvature of the phase transfer function of the BM, at

least at medium to high frequencies (see, e.g., Smith *et al.*, 1986; Shera, 2001; Oxenham and Dau, 2001, 2004). The temporally inversed iBMTC are trains of down-chirps. A relatively narrow-band BM excitation can be expected at each point in time for a single chirp that moves apically in time, similar to the excitation of a click but moving slower in the apical direction. The spectral summation across frequency will result in a flat response as a function of time (only shaped by the spectral sensitivity of the cochlea and the frequency characteristic of ear canal and middle ear). In contrast to the stimulation with BMTC, not all filters contribute simultaneously; instead, only a few adjacent filters will

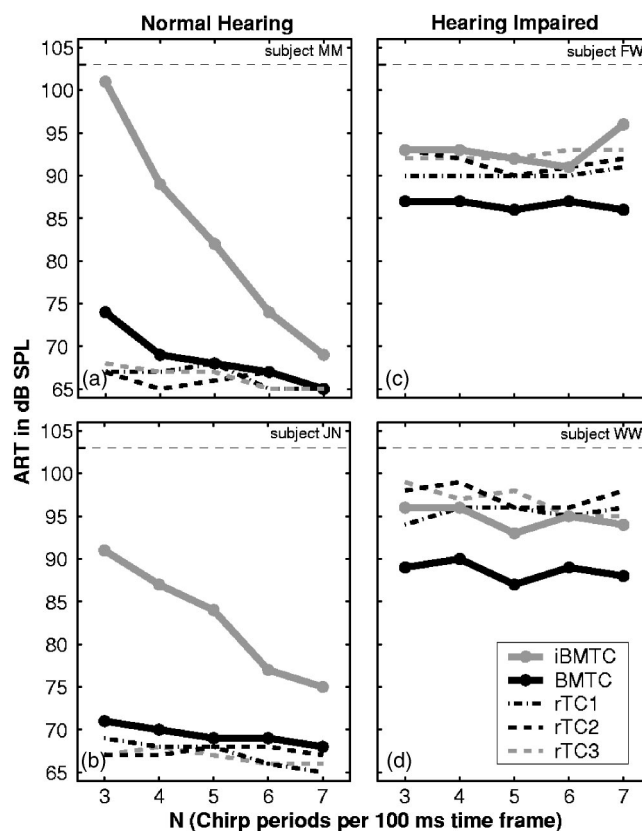


FIG. 6. Examples of ARTs measured for two individual NH [panels (a) and (b)] and two individual HI subjects [panels (c) and (d)].

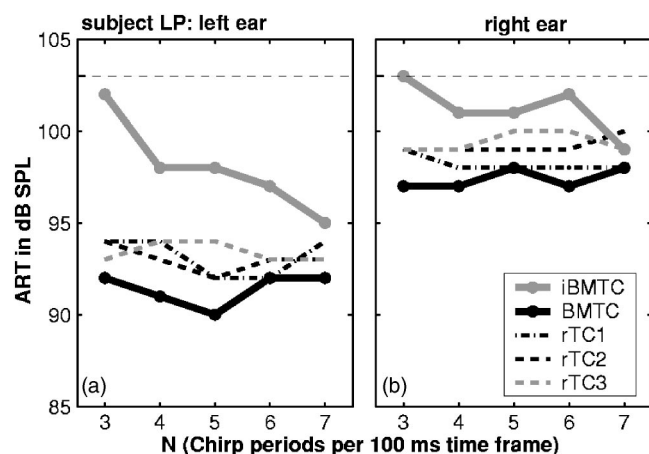


FIG. 7. Examples of the ART characteristic for different degrees of hearing loss. Subject LP showed a general difference in hearing thresholds of about 15 dB between the left (a) and right ear (b). Accordingly, the ART characteristics are different for both ears, with the better ear (a) showing attributes that can also be found in NH listeners. This indicates sufficient sensitivity to utilize this method as an indicator for the loss of BM compression that is associated with OHC damage.

contribute significantly to the spectral summation response at each point in time. From the perspective of the individual auditory filters, a relatively peaky, temporally more modulated response can be expected at the output, since the phase curvature of the down-chirps has the opposite sign as the curvature of (most of) the cochlear phase transfer functions (see, e.g., Oxenham and Dau, 2001). Finally, the rTC stimuli are tone complexes with random phases. These noise-like stimuli are expected to produce a spectrally flat response during the whole stimulation period.

## 2. Spectral summation and temporal integration

Overall, the acoustic reflex elicitation seems related to the overall spectrally summed cochlear (neural) excitation within a certain time window.<sup>7</sup> The observation that the ARTs in response to  $\text{BMTC}_3$  and  $\text{BMTC}_6$  in hearing-impaired subjects are at the same rms level allows a rough estimation of the minimal integration time constant, assuming that nonlinear effects are strongly reduced or absent in the HI subjects. The spectrally summed excitation for  $\text{BMTC}_6$  comprises two smaller peaks for every peak in the  $\text{BMTC}_3$  output signal. In order to obtain the same reflex threshold for  $\text{BMTC}_3$  and  $\text{BMTC}_6$  (as seen in the HI subjects), the temporally integrated activity/excitation must be the same for the two stimuli. This would be achieved by an integration time window of at least 30 ms, sufficient to include a full chirp of the  $\text{BMTC}_3$  stimulus and at least two peaks of the spectrally summed cochlear excitation related to two consecutive chirps of the  $\text{BMTC}_6$  stimulus. However, the criterion for reflex elicitation is not clear. A simple energy summation cannot explain the ART differences between BMTC and iBMTC stimuli. Instead, the differences could possibly be explained by assuming a peak integrator that sums up only contributions of the spectral summation response that exceed an internal threshold. The peaks in the spectral summation response of the BMTCs due to their higher synchronized excitation on the BM exceed this inter-

nal threshold at lower stimulus levels than the iBMTC and rTC with their flat temporal envelope of the spectral summation response. However, since the relative amount of excitation that is cut off by the internal threshold increases with decreasing stimulus amplitudes, and since the stimulus amplitude decreases with increasing  $N$  while keeping the rms level constant, this model would lead to the prediction of slightly increasing ARTs with  $N$ . Thus, for a more detailed model further aspects of processing have to be taken into account.

## 3. The influence of neural synchronization on the acoustic reflex

Nevertheless, it is reasonable to assume that the “gain” obtained with the BMTCs in the HI subjects, reflected in their lower ARTs relative to iBMTC and rTC stimulation [Fig. 5(b)], can be ascribed mainly to the higher neural synchronization. Similar to the explanations for the higher responses in ABR and MEG measurements using BM chirps (Dau, 2003; Rupp *et al.*, 2002), this can be explained by the higher peakiness of the spectral summation response as a function of time. Although derived from a passive BM model, BMTC or BM chirps have so far been tested only in NH subjects (e.g., Dau, 2003). It is not clear whether the improvement obtained with the BM chirp in NH subjects can be expected to hold for HI subjects. A broadening of the BM filters, i.e., a loss of tuning of the BM filters in the HI subjects, may cause a reduction in BM travel time and thus a change of the neural synchronization effect by the stimuli. In turn, it might be that the greater differences between the ARTs for BMTC and iBMTC stimuli, as observed in the NH subjects compared to the HI group [compare Figs. 5(a) and (b)], might reflect the better suitability of the stimuli for compensating the travel-time differences in the healthy cochlea.

## 4. The influence of cochlear compression on acoustic reflex thresholds

However, with increasing hearing loss the ARTs for the rTC stimuli show a stronger reduction than for the BMTC/iBMTC stimuli. This observation can hardly be explained by a change of the dispersive properties of the BM. Timing effects should not strongly affect these noise-like rTC stimuli, whereas the gain of the spectrally summed activity for the other stimuli is probably influenced strongly by the fast-acting compression in the peripheral auditory system. As is known, e.g., from models of loudness, it is generally assumed that the input from a broadband stimulus to each auditory filter is compressed separately before being summed up across frequencies. Thus, a broadband BM excitation will lead to a higher overall output in comparison to a narrow-band excitation. Zwicker and Fastl (1999) describe spectral loudness summation of up to 20 dB in NH subjects for broadband noises centered at 4 kHz, while nearly no loudness summation was found in HI subjects. Thus, the difference of loudness summation between NH and HI subjects is in the order of the gain observed here for the ARTs from the noise-like rTC stimuli in NH subjects in comparison to HI subjects. Although BMTC, iBMTC, and rTC show the same

long-term spectrum, they possess different BM excitations in time. The iBMTC is assumed to produce a high local excitation at each point in time and therefore obtain less gain (maybe almost no gain) from a fast-acting compressive nonlinearity in comparison to the excitation caused by BMTC or rTC which is spread across multiple auditory filters at each time. In each auditory filter, the iBMTCs are expected to produce the peakiest response in time, the BMTCs are assumed to show only a slightly modulated temporal excitation, and the excitation of the rTCs in each local filter will be almost flat in time as well. Assuming an almost instantaneous compression, this will lead to a further gain of the rTC from nonlinear compression in comparison to BMTC and especially iBMTC, since a series of instantaneously compressed low-amplitude excitations will result in a higher integrated output than the respective excitation with only a few higher peaks.

Another observation, the decrease of the ARTs for BMTC and iBMTC with  $N$  that can be observed in the NH subject group, also cannot be associated with a change in neural synchronization, since the phase characteristics for the single chirps are kept constant with increasing  $N$ .<sup>8</sup> Similar to the decrease in the absolute rTC thresholds, this observation might also be explained by the effects of peripheral compression. Both BMTC and iBMTC produce a temporally defined excitation in each local BM filter. Assuming a static power-law compression in the local cochlear filters, the sum of the compressed output for three excitations with a given amplitude resulting from a BMTC<sub>3</sub> elicitor would be smaller than that of six excitations evoked by a BMTC<sub>6</sub> stimulus at the same overall rms level.<sup>9</sup> Furthermore, the decrease of ARTs with  $N$  might be related to an interaction of successive stimuli on the BM. Especially for the narrow-band excitation of the iBMTC, slowly moving along the cochlear partition, it can be assumed that, for higher  $N$ , the excitation of the preceding chirp is still moving towards the apex while the excitation of the current chirp is starting at the base of the cochlea. In the case of a compressive cochlear nonlinearity, the output for lower-level inputs to many filters will exceed the output from a single filter with a respectively higher input, and thus result in lower ARTs. This effect would be more pronounced for the iBMTCs than for the BMTCs, since the iBMTCs are expected to produce a narrow-band excitation slowly moving from the base to the apex in contrast to a synchronized (already) broadband BM excitation from the BMTC. A presumed reduction in travel time on the BM for the damaged cochlear might result in a reduced effect in the HI subject group, leading to no or only a slight dependency on  $N$  for these subjects.

Overall, assuming a different gain of neural synchronization for the different stimuli in combination with a major effect of a fast-acting cochlear nonlinearity (in NH subjects) on the observed effects of ART for the different stimuli gives the qualitatively most consistent view on the data. Most HI subjects with severe hearing loss have a strongly reduced compression. In these subjects [see Figs. 6(c) and (d)], the observed effects are dominated by the gain of neural synchronization. There is no ART decrease with increasing  $N$ , and the ART from the rTC stimuli is similar to or slightly

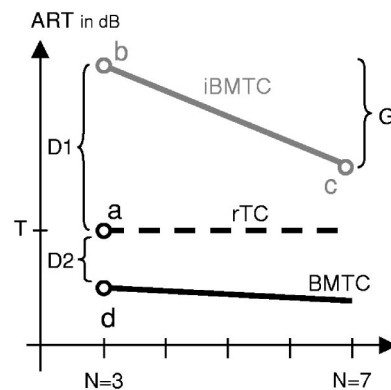


FIG. 8. Illustration of ART characteristics  $T$ ,  $D1$ ,  $D2$ ,  $G$  as given in Table I for all subjects obtained from ART measurements at only four different stimulus conditions (a,b,c,d).  $T$  is the ART for an rTC stimulus at  $N=3$  given by measurement point a.  $D1$  (b-a) is given by the ART difference for an rTC and the iBMTC stimulus at  $N=3$  and  $D2$  (d-a) as the difference for an rTC and the BMTC, respectively. Finally,  $G$  (b-c) gives the ART difference for iBMTC stimuli at  $N=3$  and  $N=7$ .

below the iBMTC stimuli. The peakier overall excitation of the BMTC might be used by a mechanism based on a peak integrator to obtain lower ARTs. With decreasing hearing loss and increasing influence of a nonlinear compression, the ARTs of the rTC stimuli are shifted towards the ARTs of the BMTC stimuli, which are also reduced [see Fig. 7(b)], and even the decrease of the ARTs with increasing  $N$  becomes observable for the iBMTC stimuli [see Fig. 7(a) and Figs. 6(b) and (a)]. Thus, besides the absolute ARTs, the differences between the ARTs for different stimuli might be used to improve the value of ART measurements as a screening tool in clinical diagnostics. For example, the difference  $G$  for ART from iBMTC stimuli at  $N=3$  and  $N=6$  or the differences  $D1$  and  $D2$  between the ARTs of iBMTC, BMTC, and rTC might be useful to indicate a loss of compression.

## B. Prediction of hearing status and clinical applications

Besides the absolute thresholds like the threshold  $T$  for the rTC (see Fig. 8, Table I), other characteristics of the ART allow for a clear distinction between NH and HI subjects, such as (1) the decrease for ART especially for iBMTC, but also for BMTC, stimulation in NH subjects with increasing  $N$  (indicated by  $G$ —see Fig. 8, Table I), while no dependency was found for the HI subjects, and (2) larger ART differences between diverse stimuli types. Thus, appropriate criteria to distinguish between NH and HI subjects may be given by the ART differences  $D1$  and  $D2$  (see Fig. 8 and Table I).  $D1$  is the difference between an rTC<sub>3</sub> and the iBMTC<sub>3</sub> stimulus, which shows a significant reduction from 22.3 dB in NH subjects to 3.5 dB in the HI subjects.<sup>10</sup> The difference  $D2$  between rTC<sub>3</sub> and BMTC<sub>3</sub> is negative for most NH subjects and is positive for most HI subjects—probably closely related to an increasing loss of compression. Additionally, the difference  $G$  between ARTs from iBMTC stimuli for  $N=3$  and  $N=7$  allows a clear distinction between NH and HI subjects (see Fig. 8 and Table I). A high sensitivity of the ART differences, e.g.,  $D1$ , with respect to hearing loss and loss of compression may be indicated by the



results from subject LP (see Fig. 7). This subject showed an almost parallel shift of hearing thresholds across frequency of about 15 dB between the two ears that is clearly reflected in different values of  $T$ ,  $D1$ ,  $D2$ , and  $G$  (see Table I). Therefore, the additional consideration of ART differences for different stimuli beside the evaluation of absolute thresholds may allow a more reliable prediction of hearing impairment than using absolute ARTs alone. Overall, based on the data from this limited group of subjects, the differences of  $D1$ ,  $D2$ ,  $G$ , or  $T$  in NH and HI subjects for the specially designed stimuli in this study may offer the opportunity to utilize the measurement of ARTs for the objective prediction of hearing loss and recruitment or for hearing-aid fitting in young or uncooperative patients.

Earlier studies showed that a close relationship of the mean ART and the uncomfortable level (UCL) might exist. However, the prediction of the UCL based on the ART measurement will be inaccurate because of the high intersubject variability (review in Olsen, 1999a; Olsen 1999b, c, Margolis and Popelka, 1975; Kawase *et al.*, 1997). As opposed to these studies where loudness and ART were compared directly, the present results suggest a comparison of the differences of ARTs for appropriate stimuli (e.g., BMTC, rTC vs iBMTC) that are differently affected by cochlear compression. Thus, the large intersubject variability might be reduced if the ART differences for special stimuli are considered, rather than the absolute thresholds alone. Based on this limited group of subjects, the derived measures  $D1$ ,  $D2$ ,  $G$ , and the ART  $T$  for the rTC stimuli give at least a set of highly significant screening indicators to distinguish between NH and HI subjects (see Table I). Further studies will have to investigate if a classification of the individual hearing loss or even a quantitative prediction can be obtained by combining the different indicators in a larger group of subjects with different shapes and types of hearing loss. Another point of interest is to find stimuli with similar properties but higher frequency specificity than the ones used here.

## VI. SUMMARY AND OUTLOOK

- (i) A clear effect of neural synchronization on acoustic reflex threshold [using the low-level reflex audiometry according to Neumann *et al.* (1996)] can be observed when comparing results obtained from BMTC and iBMTC stimuli. Therefore, the acoustic reflex threshold is strongly affected by the phase properties of the stimulus and thus by the dispersive characteristics of the cochlea. The results suggest that the ART depends on the amount of synchronized neural excitation integrated across frequency.
- (ii) The large difference in ART obtained with the different stimuli used here (BMTC, iBMTC, and rTC stimuli) as well as the large difference between normal and hearing-impaired subjects can be explained qualitatively by assuming a compressive nonlinearity as typically found in BM input-output functions of normally functioning cochleae. However, in order to obtain a more quantitative understanding of the cochlear mechanisms that contribute to elicitation of the ART, modeling work is needed in future studies.

- (iii) Besides the absolute ART values, there are several other indicators of hearing loss in our (limited) group of subjects like the differences ( $D1$  and  $D2$ ) of the acoustic-reflex thresholds for rTC<sub>3</sub> and iBMTC<sub>3</sub> or BMTC<sub>3</sub> stimuli, respectively. The clear distinctions between the two subject groups by the derived measures  $D1$ ,  $D2$ , and  $G$  in combination with absolute ART may improve the use of acoustic reflex threshold measurements as an objective predictor of a loss of cochlear compression. Further studies are required to validate these measures as a clinical tool.
- (iv) The online-analysis method might be improved in future studies by incorporating a multifrequency evaluation. This might be useful to reduce the total number of consecutive stimulus presentations and consequently in measurement time without a decrease in statistical significance. Furthermore, this approach can provide additional artifact suppression with regard to the MOC efferent reflex by considering the change in group delay across frequencies. However, utilizing more than one frequency for the evaluation corresponds with an increase of the number,  $n$ , of phase values as long as all used frequencies are equally appropriate. Therefore, no relevant difference in the detection threshold, i.e., in the sensitivity of the method, can be expected.

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<sup>1</sup>The critical value typically used for the detection of a sinusoid in noise for 16 repetitions  $R_0(16,0.001)$  is 0.63 (Mardia, 1972). Here, we use the more conservative empirically established reflex elicitation threshold value of  $R_0 = 0.8$ .

<sup>2</sup>In some subjects at higher frequencies, the ART would have been detected even at lower levels while no reflex would have been detected in other subjects at these frequencies. Around 1000 Hz appears to be a frequency region of a reliable impedance change across all subjects.

<sup>3</sup>The comparison of the iBMTC/BMTC difference for both subject groups using the Wilcoxon, Mann, and Whitney U-test for independent samples ( $U = 3.5 < 6 = U_{8,7;0.05}$ ).

<sup>4</sup>The single-sided paired-samples t-test revealed significant ART differences for both stimuli under all conditions except for the ARTs in response to BMTC<sub>6</sub> compared to BMTC<sub>7</sub>.

<sup>5</sup>Paired-samples t-tests reveal no significant differences between  $N=3$  and  $N=7$  for any of the rTC-type stimuli.

<sup>6</sup>A paired-sample t-test between the thresholds in response to iBMTC<sub>3</sub>/iBMTC<sub>7</sub> and BMTC<sub>3</sub>/BMTC<sub>7</sub> revealed no significant difference between these ART pairs. Therefore, a systematic dependency of the ARTs on  $N$  can be rejected for these stimuli in the HI subject group. However, a complete pair comparison reveals single deviations from a constant threshold across  $N$ . The ART for iBMTC<sub>3</sub>, for example, are significantly higher compared to iBMTC<sub>4,5</sub> and <sub>6</sub>. For the BMTC stimuli, the ART in response to BMTC<sub>5</sub> was significantly lower compared to BMTC<sub>6</sub> and <sub>7</sub>.

<sup>7</sup>This might be some kind of leaky integrator. However, this is subject to a more detailed modeling and will not be discussed here.

<sup>8</sup>This is different from the characteristic known for Schroeder-phase tone complexes. An increase of the repetition rate, i.e., of the fundamental frequency  $f_0$  in Schroeder-phase tone complexes, means by definition a



change in sweep rate or phase curvature as well. This is not the case for the BMTC and iBMTC. An increase of  $f_0$  results simply in an increasing number of consecutive chirps within the stimulus duration. Thus, for the BMTC/iBMTC the phase characteristic of each chirp is kept almost constant as long as  $1/f_0$  does not exceed an "effective chirp duration" of about 10 ms. For higher  $f_0$  the chirps are shifted into one another.

<sup>9</sup>For example, with an amplitude reduced by a factor of 0.71. Thus, a reduction of approximately 2 dB from  $N=3$  to  $N=6$  could be expected for an exponent of 0.3, while no effect can be expected for the HI subject assuming negligible compression and therefore an exponent close to 1.

<sup>10</sup>The comparison of the iBMTC/mean(rTC) difference for both subject groups using the Wilcoxon, Mann, and Whitney  $U$ -test for independent samples demonstrates a significant reduction of this difference in the HI listeners ( $p < 0.001$ ;  $U = 0 < 2 = U_{8;7;0.001}$ ).

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